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(71) Applicant (*for DE only*): **PHILIPS INTELLECTUAL PROPERTY & STANDARDS GMBH** [DE/DE]; Steindamm 94, 20099 Hamburg (DE).

(71) Applicant (*for all designated States except DE, US*): **KONINKLIJKE PHILIPS ELECTRONICS N.V.** [NL/NL]; Groenewoudseweg 1, NL-5621 BA Eindhoven (NL).

(72) Inventors; and

(75) Inventors/Applicants (*for US only*): **GRASS, Michael** [DE/DE]; c/o Philips Intellectual Property & Standards GmbH, Weisshausstr. 2, 52066 Aachen (DE). **KÖHLER, Thomas** [DE/DE]; c/o Philips Intellectual Property & Standards GmbH, Weisshausstr. 2, 52066 Aachen (DE).

(74) Agent: **MEYER, Michael**; Philips Intellectual Property & STandards GmbH, Weisshausstr. 2, 52066 Aachen (DE).

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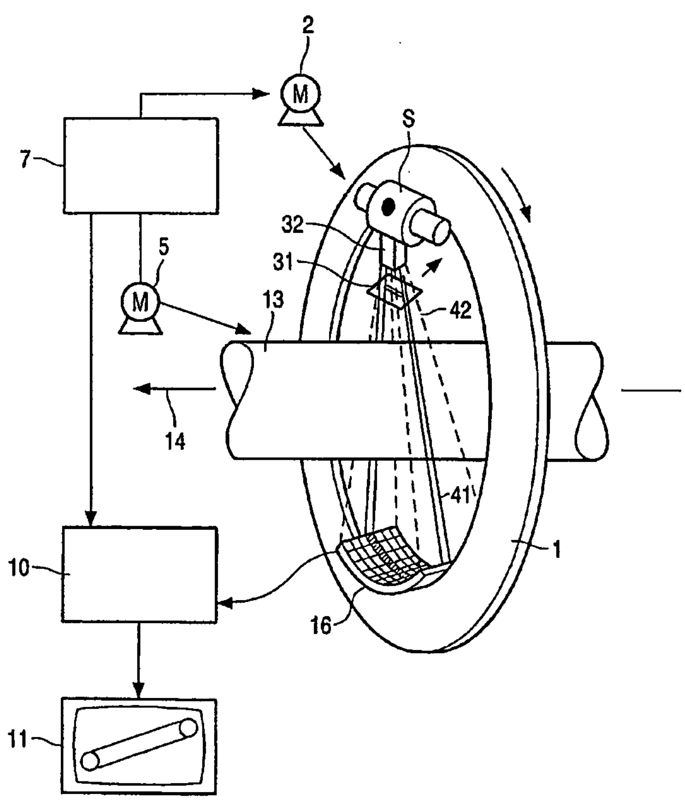
(54) Title: COMPUTED TOMOGRAPHY METHOD WITH COHERENT SCATTERED RAYS, AND COMPUTED TOMOGRAPH



(57) Abstract: The invention relates to a computed tomography method in which an examination zone is irradiated along a helix-like trajectory by a fan-like bundle of rays. In the examination zone, coherent scattered radiation is measured by a detector unit, where the spatial profile of the scattering intensity in the examination zone is reconstructed from these measured values. The reconstruction may be effected by means of a back-projection in a volume which is defined by two linearly independent vectors of the rotation plane and a wave vector transfer.

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Computed tomography method with coherent scattered rays, and computed tomograph

The invention relates to a computed tomography method in which an examination zone is irradiated by a fan-beam and a detector unit detects coherent scattered rays in the examination zone. The invention also relates to a computed tomograph for carrying out this method, and to a computer program for controlling the computed tomograph.

In known methods of the type mentioned above, the examination zone is irradiated along a circular trajectory. Coherent scattered rays are measured in the examination zone by a detector unit having a measurement surface, and the spatial profile or distribution of the scattering intensity or strength in the examination zone is reconstructed from these measured values. The reconstruction is usually effected by means of iterative methods based on algebraic reconstruction techniques (ART) or by means of two- or three-dimensional filtered back-projection. These methods are disadvantageous in that in each case only one slice of the examination zone is irradiated and subsequently also only the distribution of the scattering intensity in this one slice is reconstructed. In order to reconstruct a volume, a number of adjoining slices must be irradiated, reconstructed and combined. This takes a great deal of time and leads to imaging errors since it is almost impossible to avoid a movement in the examination zone, for example on account of the movement of a patient, during the acquisition of the measured values of two adjacent slices.

It is therefore an object of the invention to specify a method, a computed tomograph and a computer program by means of which it is possible to more rapidly acquire measured values of an irradiated volume while avoiding, or at least reducing, the abovementioned imaging errors.

With respect to the method, this object is achieved according to the invention by a computed tomography method having the steps:

a) generation of a fan beam (41) using a radiation source (S), said fan beam (41) passing through an examination zone (13) or an object located therein,

b) generation of a relative movement between the radiation source on the one hand and the examination zone or the object on the other, which movement comprises a

rotation about an axis of rotation and a displacement parallel to the axis of rotation and runs in the form of a helix,

c) acquisition of measured values which are dependent on the intensity of the radiation, by means of a detector unit (16) which detects radiation that is scattered coherently in the examination zone (13) or over the object, during the relative movement,

d) reconstruction of a distribution of the scattering intensity in the examination zone (13) from the measured values.

Compared to with known methods of the type mentioned above, the radiation source moves relative to the examination zone along a helix-like trajectory. This permits more rapid acquisition of the measured values of the entire examination zone and reduces the imaging errors that are caused, for example, by movements of the patient.

Claim 2 describes an interpolation of various measured values, which permits subsequent reconstruction in a reduced amount of time.

The scattering intensity depends not only on the material but also on the scattering angle and the wavelength of the radiation. On the other hand, the back-projection in a volume which is defined by two linearly independent vectors of the rotation plane and a wave vector transfer, as claimed in claim 3, has the advantage that the scattering intensity parameterized in this way is now only dependent on the scattering material. This is because the wave vector transfer, as is known, is proportional to the product of the inverse wavelength and the sine of half the scattering angle. The scattering angle is in this case the angle which encloses the course of the scattered ray with the course which the ray would have followed if there had not been the scattering process. In the abovementioned volume, the scattered rays have a curved shape. Taking this curved shape of the scattered rays into account in the back-projection leads to an improved quality of the reconstructed distribution of the scattering intensity.

Claims 4 and 5 in each case describe a preferred reconstruction method having computing expenditure that is lower than that of other methods and leads to a good image quality.

A computed tomograph for carrying out the method according to the invention is described in claim 6.

Claim 7 defines a computer program for controlling a computed tomograph as claimed in claim 6.

The invention will be further described with reference to examples of embodiments shown in the drawings to which, however, the invention is not restricted.

Fig. 1 shows a computed tomograph which can be used to implement the method according to the invention.

Fig. 2 shows a schematic cross-sectional view of the computed tomograph in the direction of the axis of rotation.

Fig. 3 shows a flowchart of the method according to the invention.

Fig. 4 shows a plan view of a column of detector elements.

Fig. 5 shows a schematic perspective view of a helix-like trajectory and of a slice in the examination zone.

Fig. 6 shows a schematic illustration of the arrangement of virtual radiation sources.

Fig. 7 shows a sectional zone through the rays of the virtual radiation sources.

Fig. 8 shows a schematic sectional view of the irradiated examination zone and of the detector in the direction of the rotation plane.

Fig. 9 shows the dependence of the value of the wave vector transfer on the distance of a scattering center in the examination zone from the base of the detector (point where the primary ray strikes the detector unit).

The computed tomograph shown in Fig. 1 comprises a gantry 1 which can rotate about an axis of rotation 14. For this purpose, the gantry 1 is driven by a motor 2 at a preferably constant, but adjustable, angular velocity. A radiation source S, for example an X-ray radiator, is fitted on the gantry 1. An aperture arrangement 31 determines a fan beam 41 used for the examination, said fan beam 41 being shown by solid lines in Fig. 1. The fan beam 41 runs perpendicular to the axis of rotation 14 and in the direction thereof has small dimensions, for example 1 mm. Between the aperture arrangement 31 and the radiation source S, there may be arranged a second aperture arrangement 32 which screens out a cone-shaped bundle of rays 42 from the radiation generated by the radiation source S. The cone-shaped bundle of rays 42 which would arise without aperture arrangement 31 is shown by dashed lines.

The fan beam 41 penetrates a cylindrical examination zone 13 in which an object, e.g. a patient on a patient table (both not shown) or else a technical object, may be located. After passing through the examination zone 13, the fan beam 41 strikes a detector

unit 16 fitted to the gantry 1, said detector unit 16 having a measurement surface which comprises a large number of detector elements arranged in the form of a matrix. The detector elements are arranged in rows and columns. The detector columns run parallel to the axis of rotation 14. The detector rows are located in planes perpendicular to the axis of rotation, preferably on an arc of a circle about the radiation source S. However, they may also be of a different shape, e.g. describe an arc of a circle about the axis of rotation 14 or be rectilinear. In this example of embodiment, the detector unit 16 detects not only the coherent scattered radiation but also the primary radiation and the incoherent scattered radiation.

The fan beam 41, the examination zone 13 and the detector unit 16 are adapted to one another. In a plane perpendicular to the axis of rotation 14, the dimensions of the fan beam 41 are selected such that the examination zone 13 is completely irradiated, and the length of the detector unit 16 is to be precisely dimensioned such that the fan beam 41 can be detected in its entirety. The fan beam strikes the central detector row(s).

If a technical object is concerned, rather than a patient, the object may be rotated during an examination, while the radiation source S and the detector unit 16 remain stationary. The examination zone 13 – or the object or patient table – may be displaced parallel to the axis of rotation 14 by means of a motor 5. However, the gantry 1 could also be moved in this direction in an equivalent manner.

When the motors 2 and 5 are running at the same time, the radiation source S and the detector unit 16 describe a helix-like trajectory relative to the examination zone 13. When, on the other hand, the motor 5 for displacement in the direction of the axis of rotation 14 is idle and the motor 2 is making the gantry 1 rotate, there is a circular trajectory for the radiation source S and the detector unit 16 relative to the examination zone 13. Only the helix-like trajectory will be considered in the text which follows.

Fig. 2 shows that, between the examination zone 13 and the detector unit 16, there is a collimator arrangement 6 which comprises a large number of planar lamellae 60. The lamellae 60 are made of a material that is highly absorbent for X-ray radiation and lie in planes which run parallel to the axis of rotation 14 and intersect at the focus of the radiation source S. They may be spaced apart by e.g. 1 cm, and each lamella 60 may have a dimension of e.g. 20 cm in the plane of the drawing. By means of the collimator arrangement 6, the fan beam 41 is thus divided into a number of mutually adjacent sections, so that a column of detector elements is essentially struck by only primary or scattered rays from a section.

The measured values acquired by the detector unit 16 are fed to an image processing computer 10 which is connected to the detector unit 16, for example, via a

contactless slip ring (not shown). The image processing computer 10 reconstructs the distribution of the scattering intensity in the examination zone 13 and outputs it, for example, on a monitor 11. The two motors 2 and 5, the image processing computer 10, the radiation source S and the transfer of the measured values from the detector unit 16 to the image processing computer 10 are controlled by a control unit 7.

In other embodiments, the acquired measured values may be fed for reconstruction purposes initially to one or more reconstruction computers which pass the reconstructed data to the image processing computer via a glass fiber cable for example.

Fig. 3 shows the sequence of an embodiment of a measurement and reconstruction method which can be carried out using the computed tomograph shown in Fig. 1.

Following the initialization in step 101, the gantry 1 rotates at a constant angular velocity. In step 103, the examination zone or the object or the patient table is displaced parallel to the axis of rotation and the radiation of the radiation source S is switched on, so that the detector unit 16 can detect the radiation from a large number of angular positions. The detector element(s) in the center of each detector column essentially detect the primary radiation, while the scattered radiation (secondary radiation) is detected by the detector elements lying further out in a column.

This is shown schematically in Fig. 4, which shows a plan view of a column of detector elements. The detector elements 161 which detect the scattered radiation are shown simply by hatching, while the detector element 160 in the center, which detects the primary radiation, is marked with a cross. In other radiation sources, in particular in radiation sources having a larger focus, it could also be possible for more than one detector element to detect the primary radiation. On both sides of this central detector element there are, on account of the finite dimensions of the focus of the radiation source, detector elements which are struck by scattered radiation but also by a (reduced) primary radiation. In this embodiment, only those rays which are measured by the detector elements shown by hatching in the drawing are regarded as scattered rays. The scattered rays comprise, as mentioned above, coherent and incoherent radiation, where, as is known, only the coherent radiation plays a part in the reconstruction of the distribution of the scattering intensity in the examination zone.

The scattering intensity is dependent, inter alia, on the energy of the scattered X-ray quantum. Therefore, the energy of the scattered X-ray quantum either must be measured, which requires that the detector elements are able to measure energy in a resolving manner, or else X-ray radiation with quantum energies from a range that is as small as

possible (ideally monochromatic X-ray radiation) must be used. There are various possibilities for minimizing the energy difference of the X-ray quanta relative to their energy as far as possible:

- The use of suitable filter materials, e.g. copper, in the primary ray. As a result, the soft X-ray radiation generated by an X-ray radiator, that is to say X-ray radiation having a low quantum energy, is largely suppressed.

- In addition, the voltage of an X-ray tube can be optimized relative to the selected filter.

- Finally, it is possible to make use of the so-called "balanced filter" technique. In this technique the data are acquired twice, where in each case filters having slightly differing atomic numbers are located in the ray path, the K-edge of which filters is used for filtering. The difference signal is then extracted from the two measurements.

In step 105, the measured values of the scattered rays are standardized. For this, the measured values of each radiation source position of the scattered rays are divided by the measured values of those primary rays which have caused the scattered rays.

In step 107, the measured values are reinterpolated. Here, the measured values are reinterpolated as if the examination zone had been irradiated slice by slice by rays of a radiation source which moves along a circular trajectory. This is explained in more detail with reference to Fig. 5 by way of example for a slice 21. The slice 21 is oriented perpendicular to the axis of rotation 14 and intersects the axis of rotation at the point $z = z_0$. In addition, the slice 21 intersects the helix-like trajectory 18 of the radiation source S at a point 23, so that the slice 21 alone has been irradiated from this interface position 23. Fictitious rays of the virtual radiation source which is moving along a circular path, said fictitious rays irradiating the slice 21 from other angular positions, or the corresponding measured values, are obtained by interpolation. For each projection angle of the slice 21 and for each detector element at the position $z = z_0$, the associated measured value is determined by interpolation of the next adjacent value measured in the z direction at the same projection angle and by the same detector element. In the example shown in Fig. 5, these two measured values have been acquired for a given projection angle and a given detector element at the positions $z = z_1$ and $z = z_2$. The distance $d = z_2 - z_1$ corresponds to the translational relative movement, that is to say e.g. to the advance of the table, during a complete rotation of the radiation source. In order to determine the measured value at the position $z = z_0$, the measured values at the positions $z = z_1$ and $z = z_2$ are in each case multiplied by a weight factor, said weight factor being greater the smaller the distance in the z direction of the

respective measured value from the position $z = z_0$, and added together. In this embodiment, the measured value at the position $z = z_0$ is determined by the following equation:

$$P(z_0,\alpha)=\frac{|z_0-z_2|}{|z_2-z_1|}P(z_1,\alpha)+\frac{|z_0-z_1|}{|z_2-z_1|}P(z_2,\alpha). \qquad (1)$$

Here, $P(z,\alpha)$ refers to a measured value of a detector element at the position z at the projection angle α. The interpolation is carried out for each detector element, for each projection angle and for each slice of the irradiated examination zone. In other embodiments, the interpolations could also be carried out only for all slices that are to be reconstructed.

This type of interpolation is generally known as "360° interpolation" and has been published inter alia in "Bildgebende Systeme für medizinische Diagnostik" ["Imaging systems for medical diagnosis"], H. Morneburg, Erlangen: Siemens Publicis MCD Verlag, 1995, to which reference is hereby made.

Hereinbelow, it is assumed that the examination zone has been scanned slice by slice by the above-described system of radiation source and detector, where radiation source and detector move in a circular trajectory for each slice.

In step 109, a rebinning of the interpolated measured values may take place. Here, each measured value is assigned a line from the detector element at which the measured value has been detected to the radiation source position. It is therefore assumed that rays of the fictitious, cone-shaped bundle of rays 42 have caused the measured values without the rays having been scattered. By means of the rebinning, the measured values are now resorted as if they had been measured with a different radiation source (a radiation source shaped like the arc of a circle, which can emit fans of rays that are in each case parallel to one another) and with a different detector (a planar, rectangular virtual detector).

This is explained in more detail with reference to Fig. 6. The reference 17 refers to the circular trajectory from which the radiation source irradiates the examination zone. The reference 413 refers to a fan-like bundle of rays which starts from the radiation source position S_0 and the rays of which run in a plane comprising the axis of rotation 14. It is possible to regard the cone-shaped bundle of rays, which is emitted from the radiation source position S_0, as being made up of a large number of planar fans of rays which are located in planes parallel to the axis of rotation 14, said planes intersecting one another at the radiation source position. Fig. 6 shows only a single one of these fans of rays, namely the fan beam 413.

Moreover, Fig. 6 also shows other fans of rays 411, 412 and 414, 415, which lie parallel to the fan beam 413 and in planes that lie parallel to one another and to the axis of rotation 14. The associated radiation source positions S_{-2}, S_{-1} and S_1, S_2 are assumed by the radiation source S before and after it has reached the radiation source position S_0, respectively. All rays in the fans of rays 411 to 415 have the same projection angle. The projection angle is the angle which the planes of the fans of rays enclose with a reference plane that is parallel to the axis of rotation 14.

The fans of rays 411 to 415 define a bundle of rays 410 having a tent-like shape. Figs. 6 and 7 show the sectional zone 420 which results when the bundle of rays 410 is sectioned by a plane that comprises the axis of rotation 14 and is perpendicular to the planes of the radiation fans 411 to 415. The two edges of the sectional zone 420 that intersect the axis of rotation are curved. This curvature can be attributed to the fact that the radiation source positions in the center (e.g. S_0) are further away from the sectional plane than those at the edge (e.g. S_2 or S_{-2}) and that the fans all have the same opening angle. For each group of fans of rays, therefore, a rectangular virtual detector 170 is defined in the planar sectional zone 420, the edges 171 and 172 of which detector 170 are given by the dimensions of the outer fans of rays 411 and 415, respectively, in the planar sectional zone.

Fig. 7 also shows – marked by round dots – the penetration points of some rays contained in the fans of rays 411…415 through this virtual detector. Finally, crosses show the support points of a regular Cartesian grid. The penetration points and the support points never coincide. The measured values at the equidistant support points within the virtual detector 170 must therefore be determined from the measured values for the penetration points.

This rebinning is described in detail in DE 198 45 133 A1, which is hereby incorporated by way of reference in the present text.

In step 111, a one-dimensional filtering using a ramp-like transmission factor increasing with the spatial frequency is applied to the measured values resulting from the rebinning. In each case successive values in a direction parallel to the rotation plane, that is to say along a row of the virtual detector, are used for this purpose. This filtering is carried out along each row of the virtual detector for all projection angles.

In other embodiments, the rebinning could be omitted. In such a case it is known to modify the filtering as the detector unit is curved e.g. in an arc-shaped manner about the radiation source or about the axis of rotation.

The filtered measured values are then used to reconstruct the distribution of the scattering intensity in the examination zone by means of back-projection.

The back-projection in this case takes place in a volume which is defined by the vectors $\vec{x}$, $\vec{y}$ and $\vec{q}$, where the unit vectors $\vec{x}$ and $\vec{y}$ lie in the rotation plane and are oriented horizontally or vertically and the wave vector transfer $\vec{q}$ is oriented parallel to the axis of rotation. In other examples of embodiments two different, linearly independent vectors of the rotation plane may be used instead of the vectors $\vec{x}$ and $\vec{y}$. The value of the wave vector transfer $\vec{q}$ is, as already mentioned above, proportional to the product of the inverse wavelength λ of the scattered X-ray quanta and of the sine of half the scattering angle Θ:

$$q = (1/\lambda)\sin(\Theta/2) \qquad (2)$$

The scattering angle Θ can be determined using the arrangement comprising the examination zone 15 irradiated by the fan beam 41 and the detector unit 16, which arrangement is shown in Fig. 7. A detector element D_i detects scattered rays which have been scattered at various scattering angles Θ. These scattering angles Θ can be calculated in accordance with the following equation:

$$\Theta = \arctan(a/d) \qquad (3)$$

Here, d is the distance of a scattering center S_i and a is the distance of the detector element D_i from the base 12 of the detector.

The detector element D_i detects rays which are scattered at the angles $\Theta_1<\Theta<\Theta_2$ in the examination zone 15 irradiated by the fan beam 41.

From the two equations mentioned above there results, for small angles Θ:

$$q \approx a/(2d\lambda) \qquad (4)$$

The value of the wave vector transfer $\vec{q}$ as a function of the distance d of a scattering center from the base of the detector thus has a hyperbolic and thus nonlinear course (shown in Fig. 8). From this it can be seen that the originally straight course of the scattered

rays is curved in the (x,y,q) space. The back-projection is thus effected along rays curved in a hyperbolic manner.

In step 113, a voxel V(x,y,q) is determined within a predefinable (x,y) region (field of view – FOV) and within a value range of the wave vector transfer $\vec{q}$ resulting from the geometry of the computed tomograph.

In step 115, the filtered values are multiplied by a weight factor which corresponds to the reciprocal value of the cosine of the scattering angle. As a result, the decreasing effective detector zone as the scattering angle increases is taken into account. If said angle is small, the cosine of the angle is practically always 1, so that this weighting can be omitted. In addition, account is taken of the decreasing ray density as the distance of the scattering center, i.e. of the voxel V(x,y,q), from the point where the scattered ray strikes the measurement surface increases, by all measured values for each radiation source position being multiplied by a weight factor which corresponds to the square of the distance between the scattering center at which the ray associated with the measured value was scattered and the point where the scattered ray strikes the measurement surface.

If in other embodiments the rebinning is omitted, then an additional multiplication of the filtered measured values by a weight factor is necessary, said weight factor corresponding to the reciprocal value of the square of the distance between the radiation source position and the scattering center at which the detected ray was scattered.

In the back-projection in step 117, all curved rays which match the voxel V(x,y,q) are now used. If no ray of a radiation source position exactly matches the center of the voxel, then the associated value must be determined by interpolation of the measured values of adjacent rays. The measured value which can be assigned to the ray matching the voxel or the measured value obtained by interpolation is accumulated at the voxel V(x,y,q). Once the values for the relevant voxel have been accumulated in this way for all radiation source positions, in step 119 a check is made as to whether all voxels are irradiated in the (x,y,q) zone which is to be reconstructed. If this is not the case, the flowchart branches to step 113. Otherwise, the distribution of the scattering intensity has been determined for all voxels in the FOV and the reconstruction method is terminated (step 121).

In other embodiments the back-projection can be carried out in the volume defined by two linearly independent vectors of the rotation plane and by the wave vector transfer, as an approximation along straight rays.

Moreover, in other embodiments the distribution of the scattering intensity in the examination zone can be reconstructed from the interpolated measured values by known

iterative methods based on algebraic reconstruction techniques (ART) or by known two-dimensional back-projections.

In other embodiments, initially the projections for various projection angles of a slice can be obtained by means of interpolation. This slice is then reconstructed in accordance with one of the abovementioned methods. Projections for various projection angles of another slice, e.g. an adjacent slice, are then generated by means of interpolation. This second slice is also reconstructed in accordance with one of the abovementioned methods. In this way the distribution of the scattering intensity in the entire examination zone can be reconstructed slice by slice.

LIST OF REFERENCES:

S	radiation source
$S_{-2}...S_2$	radiation source positions
1	gantry
2, 5	motor (drive arrangement)
6	collimator arrangement
7	control unit
10	image processing computer (reconstruction unit)
11	monitor
12	base of the detector
13	examination zone
14	axis of rotation
15	irradiated examination zone
16	detector unit
17	circular trajectory
18	helix-like trajectory
21	slice of the examination zone
23	interface
31, 32	aperture arrangement
41	fan beam
42	bundle of rays
60	lamellae
160	detector elements which detect primary radiation
161	detector elements which detect scattered radiation
170	virtual detector
171, 172	edges of the virtual detector
410	bundle of rays
411...415	fictitious fan beam
420	sectional zone

CLAIMS:

1. A computed tomography method having the steps:

a) generation of a fan beam (41) using a radiation source (S), said fan beam (41) passing through an examination zone (13) or an object located therein,

b) generation of a relative movement between the radiation source on the one hand and the examination zone or the object on the other, which movement comprises a rotation about an axis of rotation and a displacement parallel to the axis of rotation and runs in the form of a helix,

c) acquisition of measured values which are dependent on the intensity of the radiation, by means of a detector unit (16) which detects radiation that is scattered coherently in the examination zone (13) or over the object, during the relative movement,

d) reconstruction of a distribution of the scattering intensity in the examination zone (13) from the measured values.

2. A computed tomography method as claimed in claim 1, wherein, in the reconstruction step d), measured values are generated for each slice in the examination zone which is oriented perpendicular to the axis of rotation, for different projection angles and for each detector element by interpolating measured values from various slices, and the distribution of the scattering intensity in the examination zone (13) is reconstructed from the measured values generated.

3. A computed tomography method as claimed in claim 1, wherein, in the reconstruction step d), measured values are generated for each slice in the examination zone which is oriented perpendicular to the axis of rotation, for different projection angles and for each detector element by interpolating measured values from various slices, and a back-projection is carried out along rays of curved shape in a volume which is defined by two linearly independent vectors of the rotation plane and a wave vector transfer.

4. A computed tomography method as claimed in claim 1, wherein the reconstruction step d) comprises the following steps:

- generation of measured values for each slice in the examination zone which is oriented perpendicular to the axis of rotation, for different projection angles and for each detector element by interpolating measured values from various slices,

- rebinning of the interpolated measured values to form a number of groups, where each measured value corresponding to a detector element and a radiation source position is assigned a straight line from the detector element to the radiation source position and each group comprises a number of planes that are parallel to one another and to the axis of rotation, in which planes there is in each case one fan of lines (411...415),
- one-dimensional filtering of the measured values in a direction parallel to the rotation plane,
- reconstruction of the distribution of the scattering intensity from the measured values by back-projection.

5. A computed tomography method as claimed in claim 1, wherein the reconstruction step d) comprises the following steps:
- generation of measured values for each slice in the examination zone which is oriented perpendicular to the axis of rotation, for different projection angles and for each detector element by interpolating measured values from various slices,
- rebinning of the interpolated measured values to form a number of groups, where each measured value corresponding to a detector element and a radiation source position is assigned a straight line from the detector element to the radiation source position and each group comprises a number of planes that are parallel to one another and to the axis of rotation, in which planes there is in each case one fan of lines (411...415),
- one-dimensional filtering of the measured values in a direction parallel to the rotation plane,
- reconstruction of the distribution of the scattering intensity from the measured values by back-projection, where the back-projection is carried out along rays of curved shape in a volume which is defined by two linearly independent vectors of the rotation plane and a wave vector transfer.

6. A computed tomograph, in particular for carrying out the method as claimed in claim 1, having
- a radiation source (S) and an aperture arrangement (31) which is located between the examination zone (13) and the radiation source (S), for generating a fan beam (41) that passes through an examination zone (13) or an object located therein,

- a drive arrangement (2, 5) for making an object located in the examination zone (13) and a radiation source (S) rotate relative to one another about an axis of rotation (14) and move parallel to the axis of rotation (14),

- a detector unit (16) coupled to the radiation source (S), for acquiring measured values,

- a reconstruction unit (10) for reconstructing the distribution of the scattering intensity within the examination zone from the measured values acquired by the detector unit (16),

- a control unit (7) for controlling the radiation source (S), the detector unit (16), the drive arrangement (2, 5) and the reconstruction unit (10) in accordance with the following steps:

a) generation of a fan beam (41) using a radiation source (S), said fan beam (41) passing through an examination zone (13) or an object located therein,

b) generation of a relative movement between the radiation source on the one hand and the examination zone or the object on the other, which movement comprises a rotation about an axis of rotation and a displacement parallel to the axis of rotation and runs in the form of a helix,

c) acquisition of measured values which are dependent on the intensity of the radiation, by means of a detector unit (16) which detects radiation that is scattered coherently in the examination zone (13) or over the object, during the relative movement,

d) reconstruction of a distribution of the scattering intensity in the examination zone (13) from the measured values.

7. A computer program for a control unit (7) for controlling a radiation source (S), an aperture arrangement (31), a detector unit (16), a drive arrangement (2, 5) and a reconstruction unit (10) of a computed tomograph for carrying out the method as claimed in claim 1 in accordance with the following sequence:

a) generation of a fan beam (41) using a radiation source (S), said fan beam (41) passing through an examination zone (13) or an object located therein,

b) generation of a relative movement between the radiation source on the one hand and the examination zone or the object on the other, which movement comprises a rotation about an axis of rotation and a displacement parallel to the axis of rotation and runs in the form of a helix,

c) acquisition of measured values which are dependent on the intensity of the radiation, by means of a detector unit (16) which detects radiation that is scattered coherently in the examination zone (13) or over the object, during the relative movement,

d) reconstruction of a distribution of the scattering intensity in the examination zone (13) from the measured values.

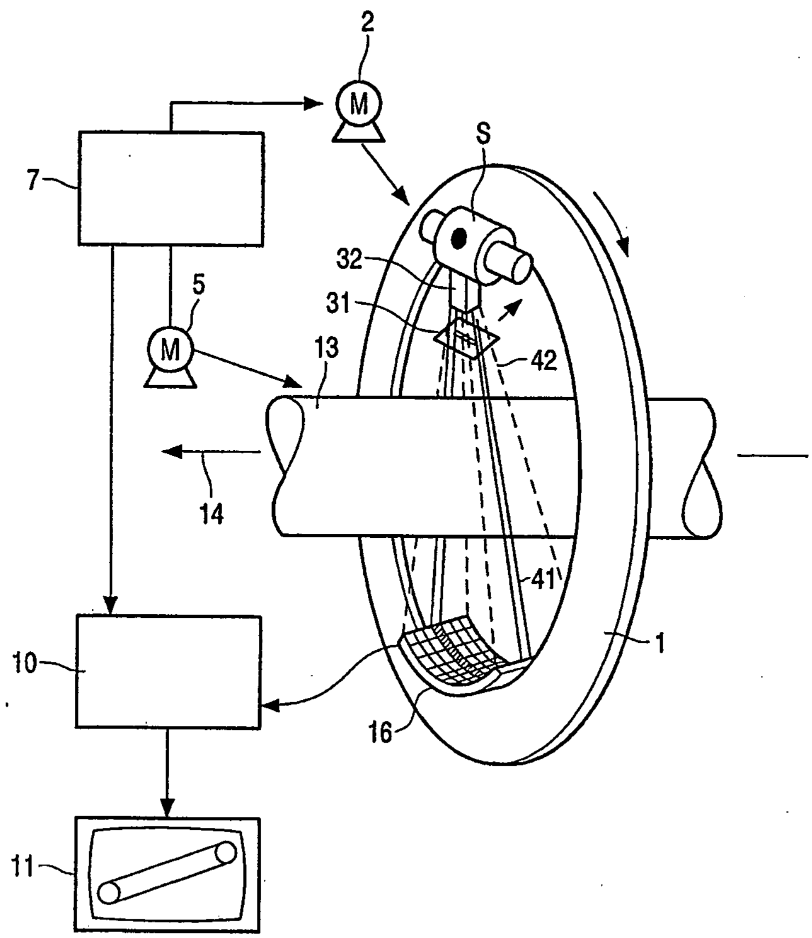

FIG.1

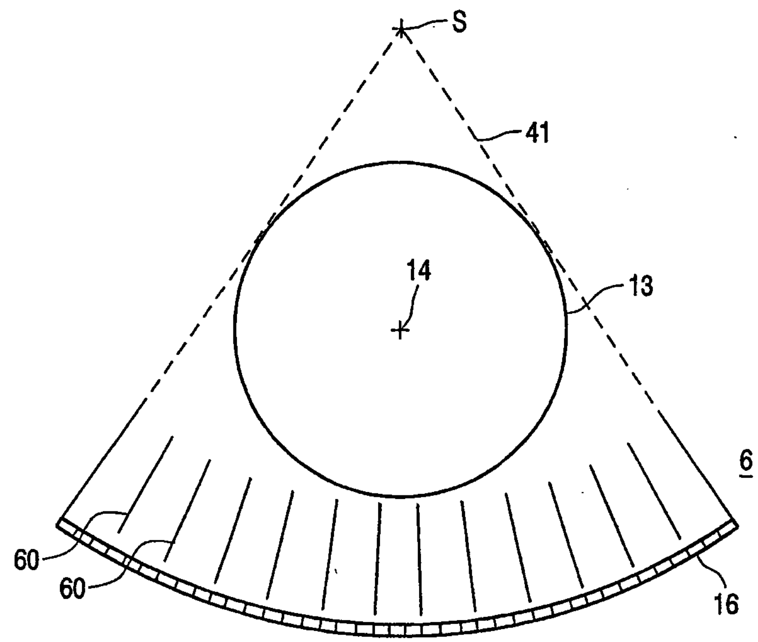

FIG.2

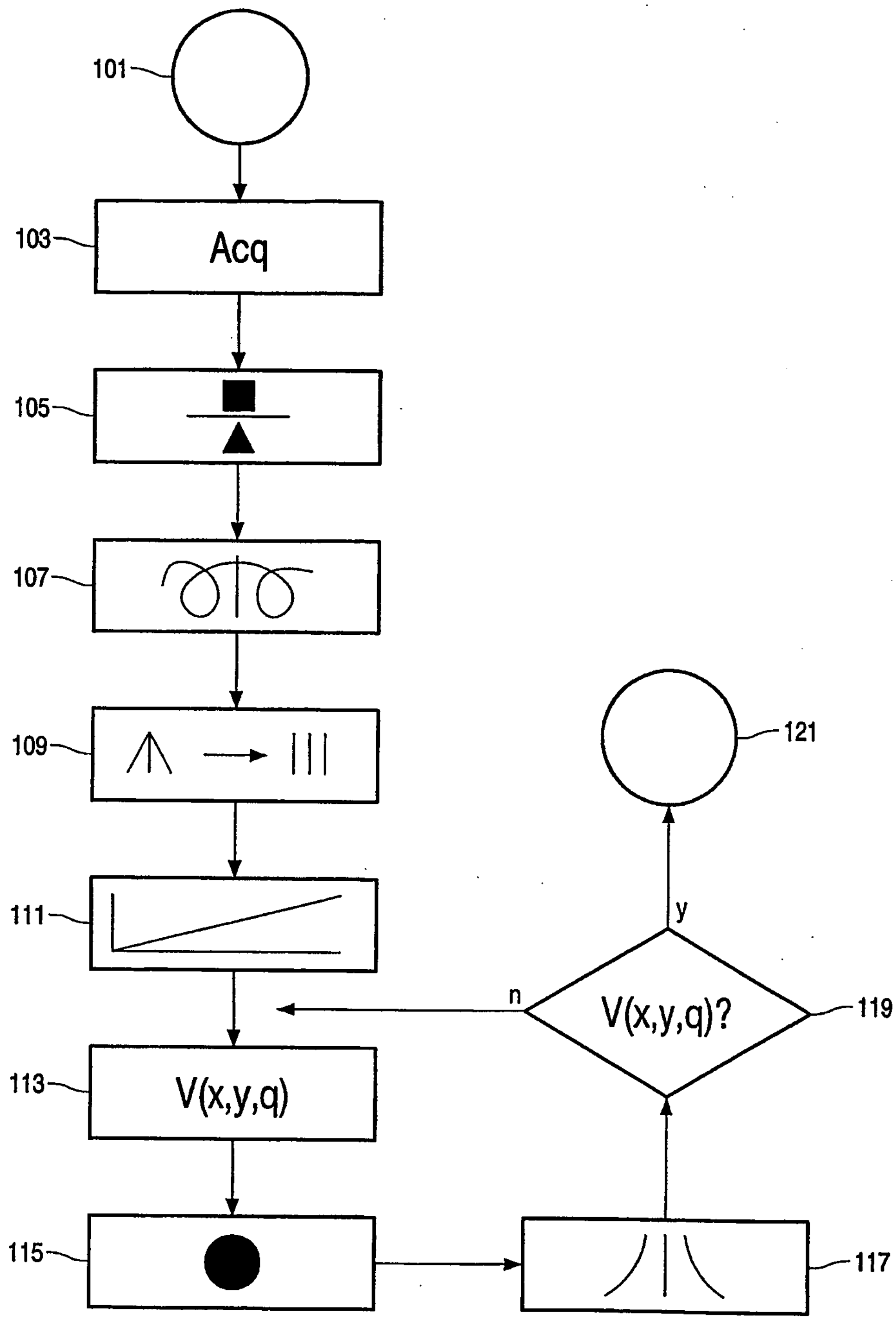

FIG.3

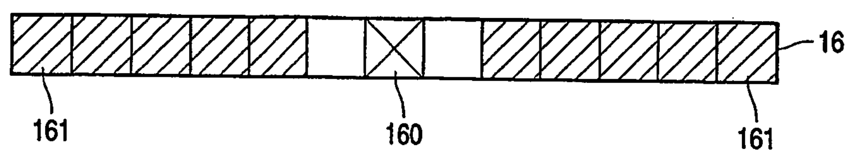

FIG.4

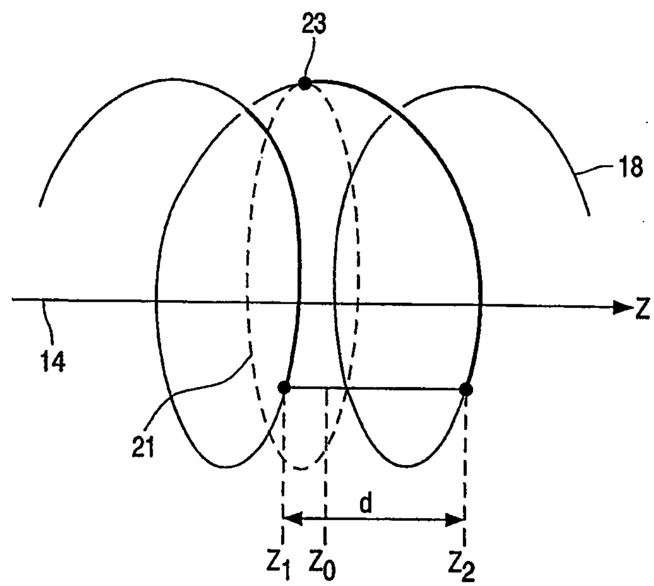

FIG.5

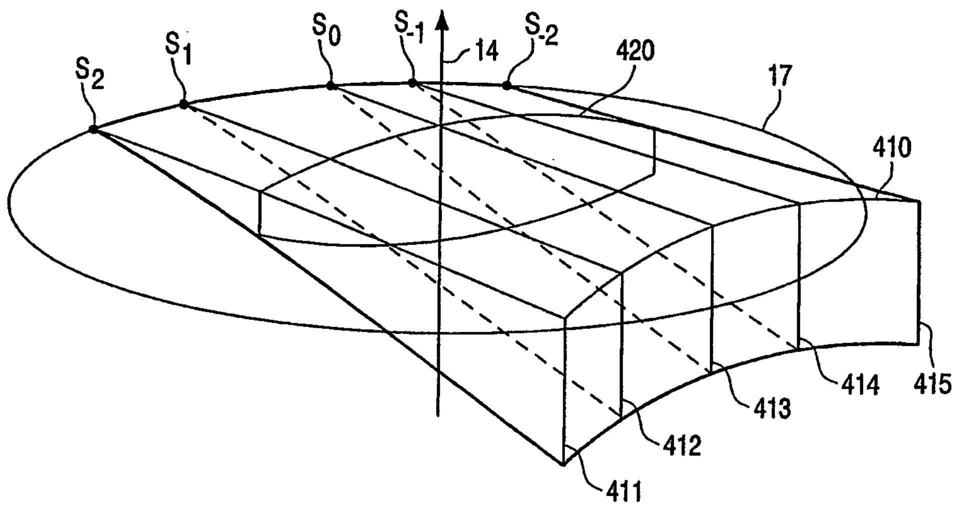

FIG.6

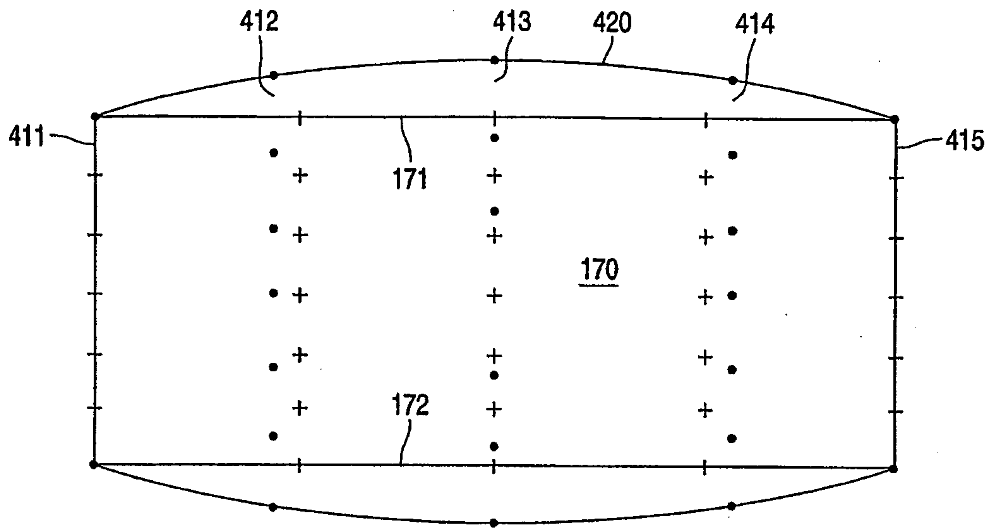

FIG.7

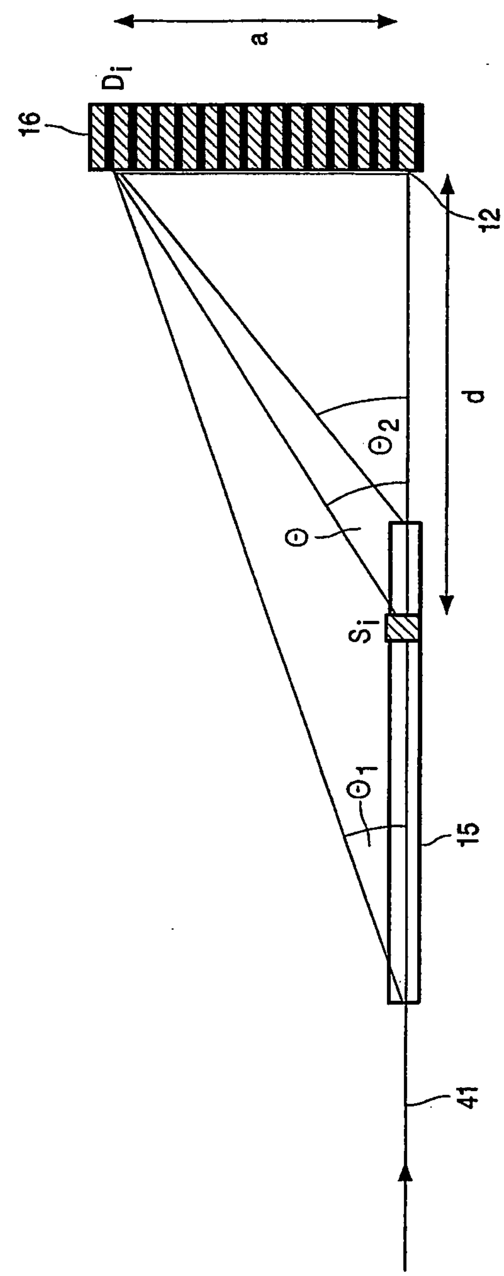

FIG.8

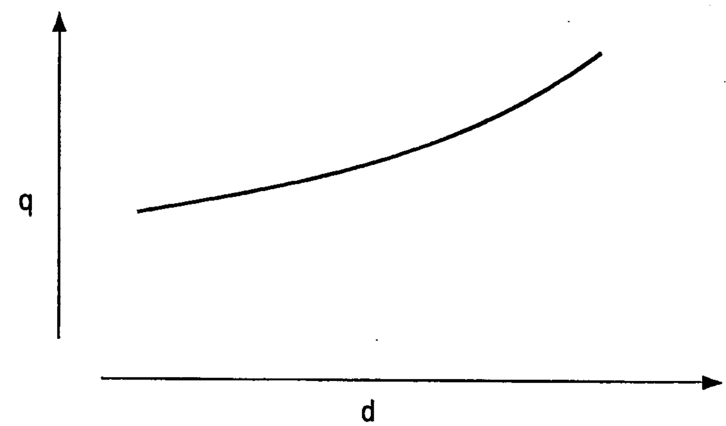

FIG.9

INTERNATIONAL SEARCH REPORT

PCT/IB 04/00110

A. CLASSIFICATION OF SUBJECT MATTER
IPC 7 G06T11/00 A61B6/03

According to International Patent Classification (IPC) or to both national classification and IPC

B. FIELDS SEARCHED

Minimum documentation searched (classification system followed by classification symbols)
IPC 7 G06T A61B

Documentation searched other than minimum documentation to the extent that such documents are included in the fields searched

Electronic data base consulted during the international search (name of data base and, where practical, search terms used)

EPO-Internal, WPI Data, INSPEC

C. DOCUMENTS CONSIDERED TO BE RELEVANT

Category °	Citation of document, with indication, where appropriate, of the relevant passages	Relevant to claim No.
X Y	US 6 470 067 B1 (HARDING GEOFFREY) 22 October 2002 (2002-10-22) abstract column 3, line 29 - line 46 column 3, line 57 - line 63 ---	1,6,7 2-5
Y	US 6 285 733 B1 (GRASS MICHAEL ET AL) 4 September 2001 (2001-09-04) cited in the application abstract column 7, line 15 - line 19 ---	2,4

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[X] Further documents are listed in the continuation of box C.

[X] Patent family members are listed in annex.

° Special categories of cited documents :

"A" document defining the general state of the art which is not considered to be of particular relevance

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"X" document of particular relevance; the claimed invention cannot be considered novel or cannot be considered to involve an inventive step when the document is taken alone

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"&" document member of the same patent family

Date of the actual completion of the international search	Date of mailing of the international search report
7 May 2004	21/05/2004

Name and mailing address of the ISA	Authorized officer
European Patent Office, P.B. 5818 Patentlaan 2 NL – 2280 HV Rijswijk Tel. (+31–70) 340–2040, Tx. 31 651 epo nl, Fax: (+31–70) 340–3016	González Arias, P

C.(Continuation) DOCUMENTS CONSIDERED TO BE RELEVANT

Category *	Citation of document, with indication, where appropriate, of the relevant passages	Relevant to claim No.
Y	D.G. BUCKNALL, S. LANGRIDGE, R.M. DALGLIESH: "CRISP Instrument Manual" 'Online! January 2002 (2002-01) XP002279154 Retrieved from the Internet: <URL: www.isis.rl.ac.uk/LargeScale/CRISP/documents/manual/crisp_manual.pdf> 'retrieved on 2004-05-03! page 6 -page 8 figure 2	3,5
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